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A Powered Lower Limb Orthosis for Providing Legged Mobility in Paraplegic Individuals

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Abstract

This paper presents preliminary results on the development of a powered lower limb orthosis intended to provide legged mobility (with the use of a stability aid, such as forearm crutches) to paraplegic individuals. The orthosis contains electric motors at both hip and both knee joints, which in conjunction with ankle-foot orthoses, provides appropriate joint kinematics for legged locomotion. The paper describes the orthosis and the nature of the controller that enables the SCI patient to command the device, and presents data from preliminary trials that indicate the efficacy of the orthosis and controller in providing legged mobility.

Keywords

Orthotic devices; robotics; rehabilitation; spinal cord injuries; walking; mobility limitation

1 Introduction

One of the most significant impairments resulting from paraplegia is the loss of mobility, particularly given the relatively young age at which such injuries occur ^{1–3}. In addition to impaired mobility, the inability to stand and walk entails severe physiological effects, including muscular atrophy, loss of bone mineral content, frequent skin breakdown problems, increased incidence of urinary tract infection, muscle spasticity, impaired lymphatic and vascular circulation, impaired digestive operation, and reduced respiratory and cardiovascular capacities ⁴.

In an effort to restore some degree of legged mobility to individuals with paraplegia, several passive and powered lower limb orthoses have been developed and described in the engineering literature. The simplest form of passive orthotics are long-leg braces that incorporate a pair of ankle-foot orthoses (AFOs) to provide support at the ankles, which are rigidly coupled to leg braces that lock the knee joints against flexion. The hips are typically stabilized by the tension in the ligaments and musculature on the anterior aspect of the pelvis. Since almost all energy for movement is provided by the upper body, these (passive) orthoses require considerable upper body strength and a high level of physical exertion, and provide very slow walking speeds. A more sophisticated orthosis, the hip guidance orthosis (HGO), is described in ⁵. The HGO incorporates hip joints that rigidly resist hip adduction and abduction, and rigid shoe plates that provide increased center of gravity elevation at toe-off, thus enabling a greater degree of forward progression per stride. Another variation on the long-leg orthosis, the reciprocating gait orthosis (RGO), incorporates a kinematic constraint that links hip flexion of one leg with hip extension of the other, typically by means of a push-pull cable assembly. As with other passive orthoses, the paraplegic individual leans forward against the stability aid, utilizing gravity to provide hip extension

of the stance leg. Since motion of the hip joints is reciprocally coupled, the gravity-induced hip extension also provides contralateral hip flexion (of the swing leg), such that the stride length of gait is increased. Examples of this type of orthosis, and studies of its efficacy, are described in ^{6–13}.

In order to decrease the high level of exertion associated with passive orthoses, some researchers have investigated the use of powered orthoses, which incorporate actuators to assist with locomotion. Historical efforts to develop powered orthoses to aid in paraplegic mobility include ^{14–16}. More recently, Ruthenberg ¹⁷ developed a powered orthosis for evaluating design requirements for paraplegic gait assistance. In ^{18–20}, a powered orthosis was developed by combining three electric motors with an RGO, two of which were located at the knee joints to enable knee flexion and extension during swing, and one of which assisted the hip coupling, which in essence assisted both stance hip extension and contralateral swing hip flexion. The orthosis was shown to increase gait speed and decrease compensatory motions, relative to walking without powered assistance.

In ^{21–24}, the authors describe control methods for providing assistive maneuvers (sit-to-stand, stand-to-sit, and walking) to paraplegic individuals with the powered lower limb orthosis HAL, which is an emerging commercial device with (in the incarnation utilized in the aforementioned publications) six electric motors (i.e., powered sagittal plane hip, knee, and ankle joints). Like the powered lower limb orthosis HAL, two additional emerging commercial devices are the ReWalk powered orthosis (Argo Medical Technologies) and the eLEGS powered orthosis (Berkeley Bionics). At the time of writing of this paper, studies have not been published regarding the design, performance characterization, or efficacy of these systems.

The authors of this paper have developed a powered lower limb orthosis to enable, with the use of a stability aid (e.g., forearm crutches), legged locomotion in paraplegic patients. This paper describes the orthosis and its controller, and presents data from preliminary trials on a single paraplegic subject (within parallel bars) that indicates the device can effectively provide legged mobility.

2 Powered Orthosis Prototype

The lower limb powered orthosis is shown in Fig. 1. The user is expected to have sufficient upper body strength to maintain balance with a walker or forearm crutches. The orthosis incorporates powered hip and knee joints. A hip cradle surrounds the lumbar and abdominal areas, enabling the orthosis to impose a hip joint torque, and providing support to the upper body. The thigh pieces are connected to the hip cradle by the hip joints and each thigh is connected to the respective shank piece through the knee joint. The four articulations (right and left hip and knee joints) provide torque in the sagittal plane while restricting rotation in the coronal and transverse planes. The hip and knee joints are powered by two brushless DC motors through a transmission, which enables each joint to provide up to 12 Nm of torque continuously and 40 Nm of torque for shorter (i.e., 2-sec) durations. As a safety measure, both knee joints include normally locked brakes, in order to preclude knee buckling in the event of a power failure. The system is designed to be used in conjunction with a standard ankle foot orthosis (AFO) to provide stability at the ankle, and to preclude foot drop during the swing phase of gait.

The orthosis is controlled by on-board electronics, with components distributed in the hip and both thigh pieces. In order to expedite control development and facilitate data collection and analysis, high-level control is implemented on a host computer (PC). The on-board electronics are powered by a lithium polymer battery, which is located in the lumbar area of the hip cradle (see Fig. 1). The primary functional elements of the electronics include data

processing (and control), sensor interface, power electronics, communication, and power management. Specifically, the data processing element consists primarily of a 32-bit microcontroller that operates at 80 MHz and contains 512 kB flash memory and 32 kB RAM. The sensor interface includes appropriate interface for the Hall effect sensors (for control of the brushless motors), potentiometer interfaces for measurement of joint angles in each of the four articulated joints, and 3-axis accelerometer interface. Note that battery voltage and motor currents are also monitored via a 12-bit A/D converter. The power electronics consist of a regenerative switching servo amplifier for each brushless motor, and a spike and hold PWM driving circuit for each knee brake. The communication within the on-board electronics utilizes a serial peripheral interface (SPI) protocol, while an RS232 serial protocol is used to exchange data between a host PC and the microcontrollers. Finally, power management includes a combination of switching and linear regulators, with which the system provides various voltages (i.e., 30V, 12V, -12V, 5V, 3.3 V and 2.5V) which are required by the on-board system.

3 Powered Orthosis Control Architecture

The orthosis controller consists of a set of low-level (i.e., joint-level) controllers, which are supervised by a high-level control structure that infers intent from the user, and based on the inferred intent, provides the appropriate joint functionality. The joint-level controllers consist of variable-gain proportional-derivative (PD) feedback controllers around each joint, where at any given time, the control inputs into each controller consists of the joint angle reference, in addition to the proportional and derivative gains of the feedback controller. Note that the latter are constrained to positive values, in order to ensure stability of the feedback controllers. With this control structure, in combination with the back drivable behavior of the joint actuator and transmission units, the joints can either be controlled in a high-impedance trajectory tracking mode, or in a (relatively) low-impedance mode, by emulating physical spring-damper couples at each joint. The former can be used in finite states where it may be desirable to enforce a predetermined trajectory (e.g., during the swing phase of gait), while the latter can be used when it may be preferable not to enforce a predetermined joint trajectory, but rather provide assistive torques that facilitate movement toward a given joint equilibrium point (e.g., providing joint damping during stand-to-sit maneuver).

The joint-level controller receives trajectory commands, as well as PD gains, from the high-level-controller, which is a finite state machine (FSM) consisting of 12 states, as shown in Fig. 2. The FSM consists of two types of states – static states and transition states. The static states consist of sitting, standing, right-leg-forward (RLF) double support, and left-leg-forward (LLF) double support. The remaining 8 states, which transition between the four static states, include sit-to-stand, stand-to-sit, stand-to-walk with right half step, stand-to-walk with left half step, walk-to-stand with left half step, walk-to-stand with right half step, right step, and left step. Each state is fully defined by the combination of a set of trajectories, and a set of joint feedback gains. In general, the latter are either high or low. The set of trajectories utilized in six of the eight transition states are shown in Fig. 3. For all the trajectories shown in Fig. 3, the joint feedback gains are set high. The joint angles corresponding to the static states of RLF and LLF double support, and standing, correspond to the final angles of the trajectories shown respectively in Fig. 3. Three states remain, which are the static state of sitting, and the two transition states of sit-to-stand and stand-to-sit. The static state of sitting is defined by zero gains, and therefore the joint angles are unimportant. The transition from stand-to-sit consists of a zero proportional gain and a high derivative gain (i.e., by damping without any stiffness). In this manner, the joint angles are also unimportant, assuming they are constant. Finally, the sit-to-stand state is defined by standing joint angles, and utilizes a set of PD gains that ramp up from zero to a value that

corresponds to a high impedance state. Table I and Fig. 3 summarize the trajectories and nature of the feedback gains that together define completely the behavior in all states of the FSM shown in Fig. 2.

The finite state controller defined by Fig. 2 and Table I is not complete without a means of appropriately switching from one to the next state, based on user commands. Such an interface should be intuitive, reliable, and controllable. The authors have developed a control interface that enables the user to control the FSM based on movements of the upper body (in conjunction with the stability aid, such as forearm crutches), as measured by sensors on the lower limb orthosis. Specifically, transitioning between states is based on the location of the (estimated) center of pressure (CoP), defined for the (assumed quasistatic user/orthosis) system as the center of mass projection onto the (assumed horizontal) ground plane. The CoP is estimated by the combination of the measured hip and knee angles (provided by joint angle sensors) and the thigh absolute orientation (provided by a three axis accelerometer). It is assumed that, with the use of the stability aid, the user can control the tilt of his or her upper body, and thus can control the location of the CoP relative to the forward foot. Thus, this distance between the CoP and the location of the forward ankle joint is utilized as the primary control input, with which the user can command all the transitions entailed in the FSM, as indicated in Fig. 2. Specifically, the transition to take the next step (right or left) is indicated by the user leaning forward (using the stability aid), such that the CoP lies within a predefined distance of the forward foot, at which point the FSM will enter either the right step or left step states, depending on which foot started forward. When transitioning from standing to walking, the user will additionally lean to one side, which indicates that the FSM should take a half step forward with the leg which is more unweighted (as detected by the 3-axis accelerometer, which is in this case utilized as a frontal plane tilt sensor). The transition from walking to standing is indicated by a pause (in double support) of sufficient time, which indicates to the FSM that the user wishes to take a half step forward into the standing static state. The transition from standing to sitting is based on a given amount of backward lean (i.e., the CoP reaches a threshold distance behind the ankle joints). Finally, transition from sitting to standing is enabled by a given amount of forward lean (i.e., the CoP reaches a threshold distance of proximity to the ankle joints). A summary of all switching conditions, governing the user interface with the FSM controller, is given in Table II.

4 Experimental Implementation

In order to provide a preliminary investigation of the ability of the orthosis and controller to provide legged mobility, the orthosis was tested on a T10 complete paraplegic subject. The subject was a 35 year-old male, 9 years post-injury, 1.85 m tall, and with a body mass of 73 kg. The experiments were conducted in parallel bars with the subject starting in a sitting position from his wheelchair. During these trials, the subject was asked to rise from sitting to standing, then walk forward to the end of the parallel bars, then transition from double-support to standing, then return to sitting in the wheelchair (which was brought to the end of the bars). Figure 4 shows representative data corresponding to such a sequence. Specifically, the sequence shown in Fig. 4 entails first rising to a standing position (transition from S1 to S2 through S5), then taking a right step (S7), then alternatively taking left and right steps (S8 and S9), returning to standing through a right step (S12), and finally sitting (S6 to S1). Note that this sequence moves through the entire state chart, with the exception of S10 and S11, which are the left-side choices that correspond to the right-side choices S7 and S12. Note that the subject, in other trials, utilizes either S10 and/or S11 in place of S7 and S12, respectively. As such, preliminary trials on this single subject indicate that measurement of the estimated CoP is sufficient to provide a control user control interface for sitting, standing, and walking movements.

5 Conclusion

The authors have developed a powered lower limb orthosis to provide legged mobility to paraplegic individuals (with sufficient upper body strength to use a stability aid), and have also described here a control structure to interface the orthosis with a paraplegic user. Specifically, by using his or her upper body control in conjunction with a stability aid, the user can affect his or her center of pressure, which is estimated by sensors on the orthosis. Preliminary testing on a single paraplegic subject (within parallel bars) indicates the orthosis and control interface offer an effective means of providing sitting, standing, and walking functionality.

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Fig. 1.
Powered orthosis prototype.
Source: Author

Fig. 2.
Finite state machine for sitting, standing, and walking.
Source: Author

Fig. 3.

Hip and knee joint angle trajectories corresponding to various states.

Source: Author

Fig. 4. Right and left leg hip and knee angle data from a walking trial, along with the corresponding state within the finite state controller.
Source: Author

TABLE I**STATE DESCRIPTION**

State	Type	Gains	Control Priority
S1-Sitting	Static	Low	NA
S2-Standing	Static	High	Position
S3-Right Forward	Static	High	Position
S4-Left Forward	Static	High	Position
S5- 1 to 2	Transition	N.A	Gain
S6- 2 to 1	Transition	N.A	Gain
S7- 2 to 3	Transition	High	Trajectory
S8- 3 to 4	Transition	High	Trajectory
S9- 4 to 3	Transition	High	Trajectory
S10- 3 to 2	Transition	High	Trajectory
S11- 2 to 4	Transition	High	Trajectory
S12- 4 to 2	Transition	High	Trajectory

Source: Author

TABLE II**FINITE STATE MACHINE SWITCHING CONDITIONS**

Transition	Condition
S1 to S5	The user leans forward and pushes himself up.
S5 to S2	The four joints meet the Standing (S2) configuration.
S2 to S7	The user leans forward and to the left.
S7 to S3	The four joints meet the Right Forward (S3) configuration.
S3 to S8	The user leans forward.
S8 to S4	The four joints meet the Left Forward (S4) configuration.
S4 to S9	The user leans forward.
S9 to S3	The four joints meet the Right Forward (S3) configuration.
S3 to S10	The user keeps the torso vertical during four seconds, then leans forward.
S10 to S2	The four joints meet the Standing (S2) configuration.
S2 to S6	The user leans backward.
S6 to S1	A preset timer is over.
S2 to S11	The user leans forward and to the right.
S11 to S4	The four joints meet the Left Forward (S4) configuration.
S4 to S12	The user keeps the torso vertical during four seconds, then leans forward.
S12 to S2	The four joints meet the Standing (S2) configuration.

Source: Author